# Sensory feedback signal derivation from afferent neurons

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# Summary of the Overall Project

In this study we are exploring the feasibility of extracting 1) cutaneous sensory information about fingertip contact and slip, and 2) proprioceptive sensory information about wrist or finger position. We use implanted nerve cuff electrodes to record peripheral nerve activity in animal models.

Our overall **objectives** for the 3-year duration of this contract are as follows:

1. Investigate, in cadaver material, implantation sites for nerve cuff electrodes from which cutaneous and proprioceptive information relevant to the human fingers, hand and forearm could be recorded.

2. Select a suitable animal preparation in which human nerve dimensions and electrode placement sites can be modeled and tested, with eventual human prosthetic applications in mind.

3. Fabricate nerve cuff electrodes suitable for these purposes, and subcontract the fabrication of nerve cuff electrodes of an alternate design.

4. Investigate the extraction of information about contact and slip from chronically recorded nerve activity using these animal models and electrodes. Specifically,

   a. Devise recording, processing and detection methods to detect contact and slip from recorded neural activity in a restrained animal;

   b. Modify these methods as needed to function in an unrestrained animal and in the presence of functional electrical stimulation (FES);

   c. Record activity for at least 6 months and track changes in neural responses over this time.

5. Supply material for histopathological examination from cuffed nerves and contralateral controls, from chronically implanted animals.

6. Investigate the possibility of extracting information about muscle force and limb position from chronically recorded neural activity.

7. Cooperate with other investigators of the Neural Prosthesis Program by collaboration and sharing of experimental findings.

# II. Summary of Progress in the Ninth Quarter

During this quarter the last of our Year Two series of implants surpassed the 180 day implant period, and we performed final acute surgeries to examine the status of the nerves and implanted devices and remove samples of the instrumented nerves. One implant was followed for 300 days. Compound action potential stability for long term chronic implants of nerve cuffs on cutaneous nerves and the stability of recorded nerve signals are discussed. Cutaneous nerve signals recorded during walking on a treadmill were used to successfully predict the simultaneously recorded muscle activity in a pilot trial of a method which could be implemented in a closed-loop FES system. Two papers accepted for publication in the proceedings of the Rehabilitation Engineering Society of North America meeting in Vancouver, Canada, in June, 1995 are included.

# III. Details of Progress in the Ninth Quarter

A.          Status of Year Two implants on Feb. 28, 1995

During the ninth quarter, we completed the periodic monitoring of compound action potentials (CAPs) to determine the status of the nerves and recording devices. The remaining two cats in the Year Two series of implants surpassed 180 days. Table 1 summarizes the status of each cat at the end of the term, detailing the total days implanted, any known problems with the nerves or implanted devices, and the final nerve CAP amplitude relative to the initial amplitude recorded at day 0. Note that at the end of the term, four cats had been terminated in a final acute (FA in Table 1) surgery according to our protocol, and that the longest implant period covered 300 days (in NIH 9).

In the Year Two series we experienced four cuff related problems: one involved nerve compression, in one cuff the electrodes deteriorated, and two involved pulled or broken wires. In NIH 9 the original proximal Median cuff induced a compression injury that caused a drastic decline in CAP amplitude. A larger diameter replacement cuff was implanted in a second surgery on day 35, and the nerve CAP recovered to 100% by the end of the implant period. In NIH 11, the proximal Median cuff showed a large increase in impedance after 180 days, and it was no longer possible to either record from or stimulate the nerve. During the final acute on NIH 11, we found that the wires to the proximal Median cuff had broken approximately 0.5 cm from the cuff, at the point of strain relief. We speculate that the wires were broken due to excessive movement of the wires leading to metal fatigue.

TABLE 1.   Year Two data summary as of Feb. 28, 1995

| Subject | Total Days implanted Final Acute | Problems with Implanted Cuffs Median | Ulnar | Radial | Final Nerve CAP Amplitude Median %, last day | Ulnar %, last day | Radial %, last day |
|---|---|---|---|---|---|---|---|
| NIH 9 | 300 FA | prox cuff replaced on day 35 due to nerve injury | | | 100%, 300 | 76%, 300 | |
| NIH 10 | 204 FA | | | | 108%, 204 | 117%, 204 | |
| NIH 11 | 281 FA | large increase in prox cuff impedance after day 180 | | | 116%, 180 | 23%, 281 | |
| NIH 12 | 289 | | | prox wires broken after day 75 | 31%, 289 | | prox wires broken after day 75 106%, 75 |
| NIH 13 | 180 FA | | | | | 87%, 180 | 89%, 180 |
| NIH 14 | 254 | | | dist wires broken on day 199 | | 132%, 254 | 70%, 191 |

The final section of Table 1 details the final CAP amplitudes and shows the relative stability of the signals over the extended implant periods. The boxes shaded in grey represent instrumented nerves that we were able to record from for a minimum of 180 days, a success rate of 11/12 or 92%. Our success rate in the First Year implant series of 8 cats was only 9/15 or 60%. The reasons for this improved longevity of implanted devices were detailed in Progress Reports #7 and #8.

While we found that the distal cuffs have shown excellent electrical and mechanical stability in implants up to 300 days as shown by impedance records, visual inspection during the final acute surgery, and the recorded CAPs detailed in Table 1, we have also found that some other devices did not survive the long term chronic implants. Shorter, larger diameter stimulation cuffs implanted in proximal locations (above the elbow) in all six Year Two cats exhibited poor recording qualities, and also suffered from electrode breakdown and large impedance increases. Large currents (5-10 mA) were required to stimulate the nerves because of the cuffs low length to diameter ratio and poor fit around the nerve. In some cats the proximal cuff could no longer be used to stimulate after some time (e.g. NIH 11). In addition, the buried EMG electrodes implanted in the forelimb muscles (Palmaris Longus, Flexor Carpi Ulnaris, Extensor Carpi Ulnaris, and Abductor Pollicis Longus) broke down after implant periods of a few months. The 631 Cooner wire used for our buried EMG electrodes was found to be not suitable for long term chronic implants or for stimulating muscles.

## B.        CAP data of Year Two implants that reached day 180

Figures 1 and 2 show the normalized latencies and amplitudes of CAP recordings in the Year Two implants for 180 days. Each of the 11 data traces (out of 12 instrumented nerves in 6 cats - the Radial nerve trace for NIH 12 ended after day 75) is interpolated to 30 day intervals from the original recording days which occured approximately every two weeks.
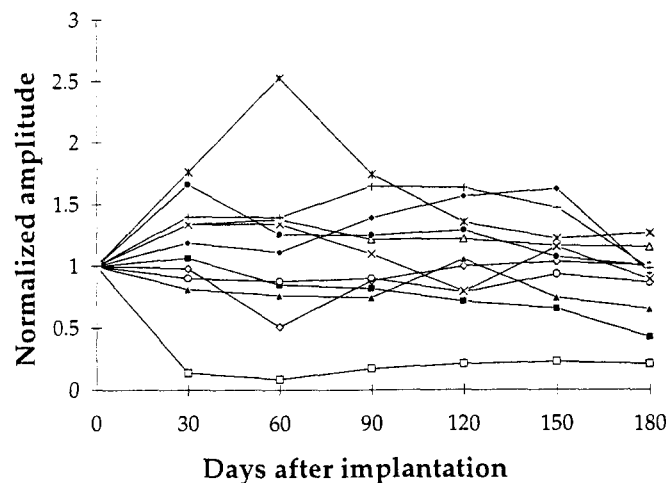


Figure 1: Normalized peak-peak amplitudes of distal CAPs
(normalized at day 0; n = 11; cats 9,10,11,12,13,14)

The Median nerve in NIH 9 that suffered a compression injury is represented by the small squares in the amplitude figure (Fig. 1), and the data show a drastic decline at around day 30 and a slow recovery up to day 180. Recordings after day 180 showed steady recovery, up to 100% at day 300. These trends are mirrored in the latency data of Fig. 2 which shows an increase in latency or conduction time at the point of injury and then a slow decline to original levels at the end of the implant period.
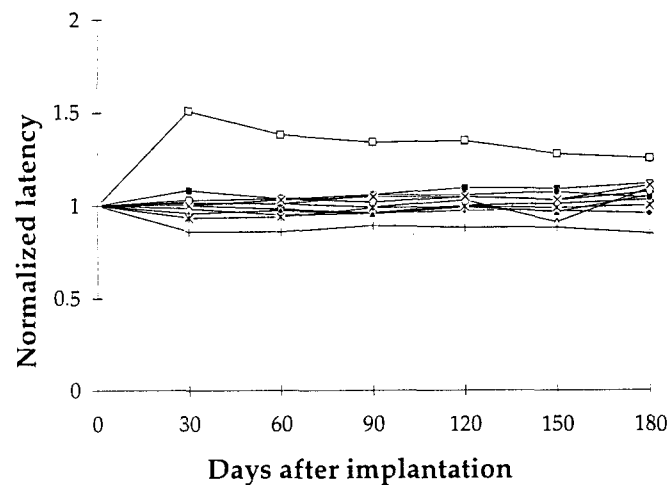


Figure 2: Normalized times to first positive peak of distal CAPs
(normalized at day 0; n = 11; cats 9,10,11,12,13,14)

Figures 3 and 4 present the average CAP amplitude and latency from 10/12 nerves in the Year Two series of implants. Two nerves that were affected by pulled wires (NIH 12) or nerve compression injuries (NIH 9) were excluded in Figs. 3 and 4. In the 10 unaffected nerves, the CAP amplitudes and latencies were very stable, with maximum standard deviations of 46% (at day 60) and 8% (at day 180) respectively.

In the Year One series of implants, we had successfully recorded from 9/15 nerves past 180 days, with 6 failures resulting from pulled wires at the percutaneous exit point (5) or damage to the nerve (1), as reported in Progress Report #5. The average amplitude and latency data had maximum standard deviations of 57% (at day 90) and 33% (at day 150) respectively.
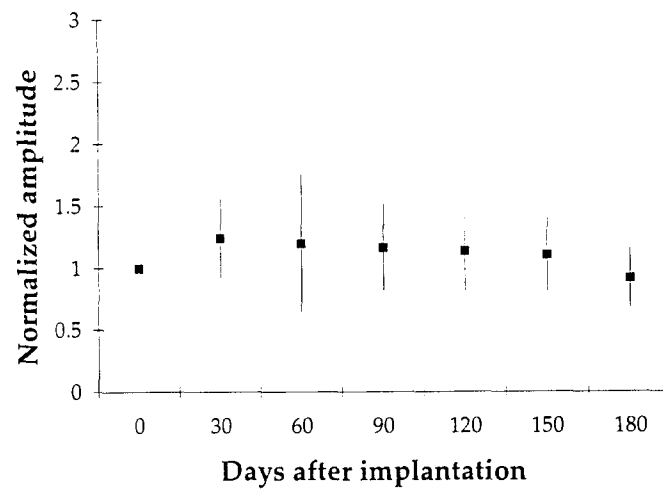
Figure 3: Average peak-peak amplitude of distal CAPs ± 1 SD
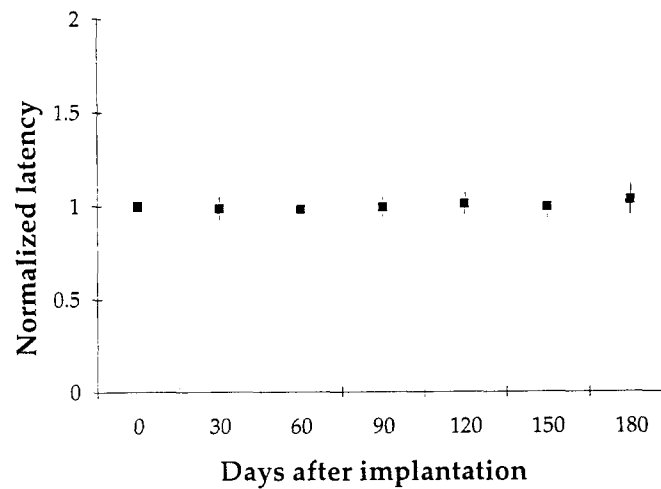(normalized at day 0; n = 10; cats 9,10,11,12,13,14)



Figure 4: Average time to first positive peak of distal CAPs ± 1 SD
(normalized at day 0; n = 10; cats 9,10,11,12,13,14)

## C.     Analysis of ENG and EMG activity patterns during walking

During the ninth quarter we analyzed some of the ENG and EMG patterns recorded during voluntary movements. In this progress report we present an example of analyzing walking data with the objective of testing the feasibility of a simple state-machine controller with ENG input to predict muscle activity as an output.

Off-line processing consisted of sampling the ENG (10 kHz) and EMG (1kHz) signals and importing six-second segments of data into Matlab. The signals were then rectified, bin-integrated to bin widths of 10 ms, and normalized to reduce the amount of data and produce relatively smooth waveforms for analysis. Thresholds for ENG and EMG signals were selected that best represented discrete events and phases in the physiological signals. A state-machine controller was designed using the binary cutaneous signals as inputs with the objective of predicting the simultaneously recorded binary muscle activity signals.

Figure 5 presents normalized data recorded from NIH 12 during walking on a treadmill. NIH 12 was instrumented with recording cuffs on the Median and Radial nerves and intramuscular recording electrodes in the Palmaris Longus (PalL), Flexor Carpi Ulnaris (FCU), Extensor Carpi Ulnaris (ECU), and Abductor Pollicis Longus (APL). The timing of the activity from the four instrumented muscles was compared to data available from the literature [1,2]: comparative data on EMG amplitude modulation from forelimb muscles during walking is not available in the literature. The flexors, PalL and FCU, exhibited strong, modulated signals corresponding closely to the stance phase of the forelimb (shown by heavy bars at bottom of Fig. 5). The extensors that we monitored showed very low levels of activity (Fig. 5 shows normalized data for convenience of analysis) and were not nearly as modulated with gait as the flexors, implying that the superficial wrist actuator muscles (flexors and extensors) were relatively passive during swing. We verified that there were no systematic differences in electrical impedances among the various implanted muscles that could have accounted for differences in levels of recorded EMGs. We have observed similar patterns of modulation in most cats in this series when walking on the treadmill.
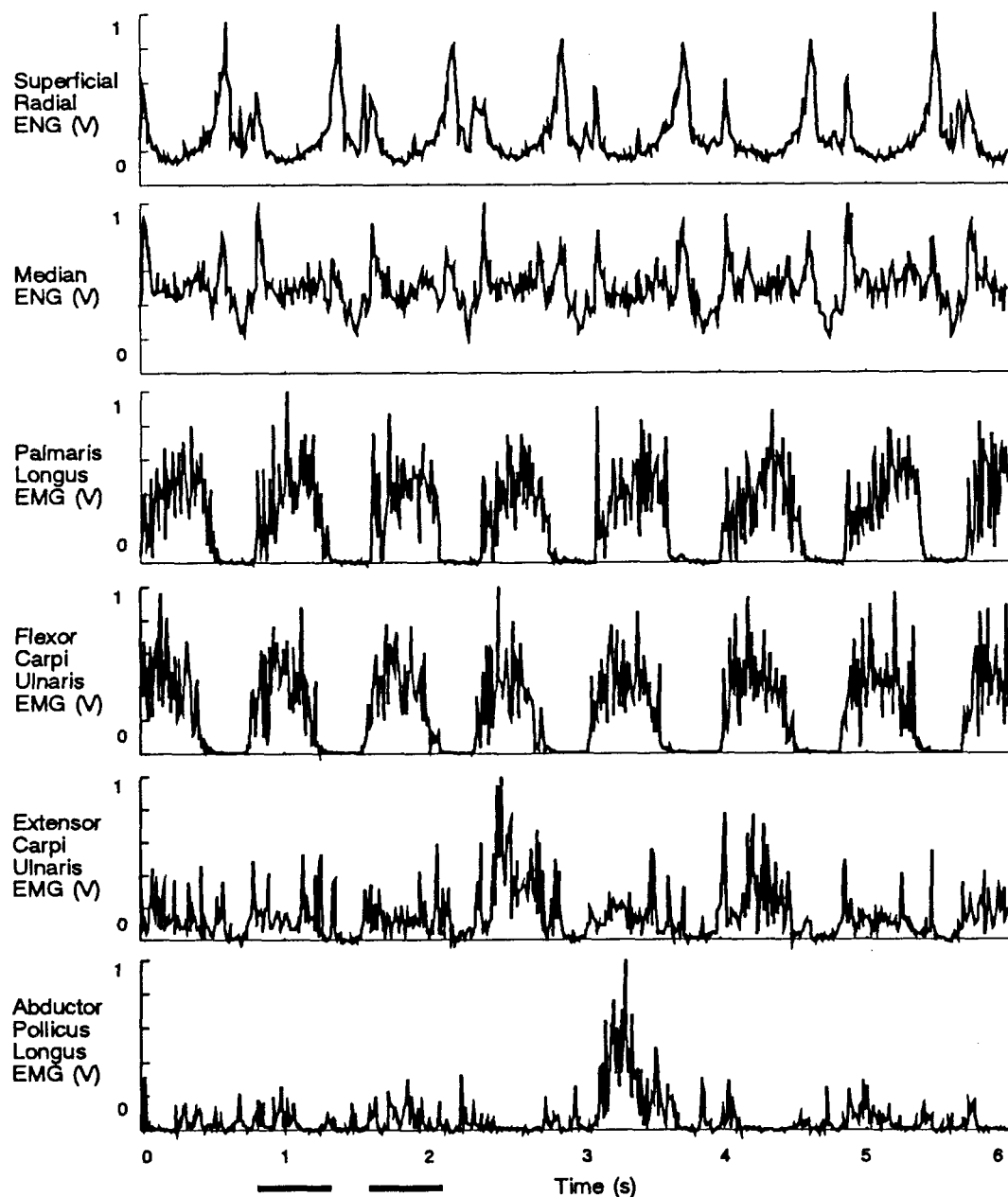


Figure 5:  ENG and EMG data from cat walking on a treadmill (NIH 12, day 85, 0.5 m/s)

Figure 6 shows processed nerve (from top: Superficial Radial, Median) and muscle (Palmaris Longus) signals with their corresponding binary levels of activity along with the binary output of the state-machine controller (bottom: predicted timing of Palmaris Longus activity). The state machine implemented two states, representing a stimulation phase and a quiet phase, and the output closely predicted the recorded timing of PalL activity. The inputs of the state machine included the current state of the controller and the two binary ENG signals, and specifically detected the large burst of Median nerve activity related to foot contact and the beginning of the stance phase (stance phase shown by heavy bars on the time scale) and the large burst of Radial nerve activity generally related to lift-off and the initiation of the swing phase. The state-machine controller also implemented a brief delay (50-100 ms) to avoid noisy transitions between states.
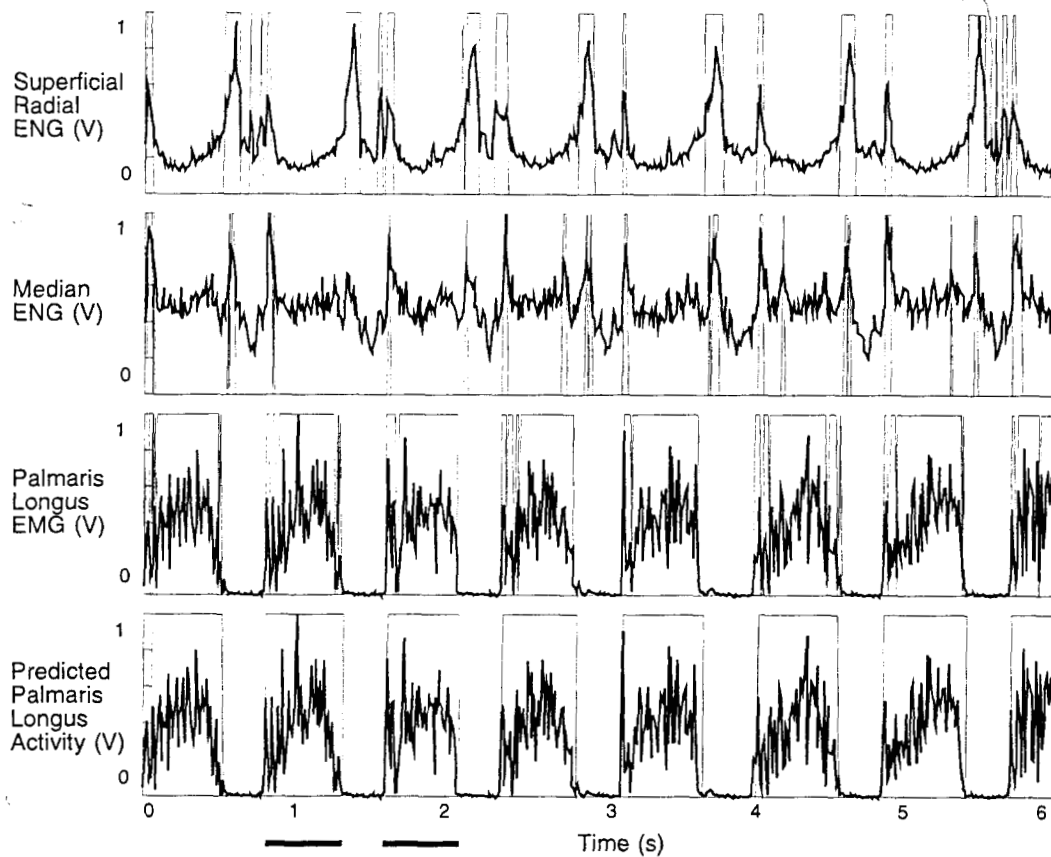


Figure 6:  ENG and EMG data from cat walking on a treadmill (NIH 12, day 85, 0.5 m/s),
and timing of PalL activity predicted by state-machine controller.

Thus, a relatively simple state-machine controller that utilizes two cutaneous ENG signals as inputs to determine state changes can accurately predict the timing of PalL activity in the cat forelimb during walking. The controller specifically looked for neural events related to paw contact and the beginning of stance and to paw lift-off and the beginning of swing, making the controller output independent of gait speed or step duration, unlike previous closed-loop FES systems that utilized counters to determine the duration of certain states such as swing [3]. This feature is of interest because state-machine controllers should be robustly designed to accommodate changes in rate of walking and noisy input signals which can lead to erroneous state changes.

More complex controllers involving pattern recognition and forms of neural networks are being developed that may theoretically accommodate a wider range of input conditions, such as rate changes, muscle fatigue, spasticity, and may accurately predict both the timing and continuously varying levels of muscle activity [4,5]. Our collaborative projects with Dr. Dejan Popovic and Zoran Nicolic in Miami and with Dr. Richard Stein and Aleks Kostov in Edmonton will centre on using more complex controllers to analyze physiologic signals and predict patterns of muscle activity.

## D.    Development of implantable nerve cuffs

During the ninth quarter we applied for  US and Canadian Patents for a novel nerve cuff design including a new opening and closing technique, and we have also applied to the Science Council of British Columbia for a grant to proceed with hiring an independent consultant to study the market of implantable devices such as nerve cuffs.  The University/Industry Liaison Office at Simon Fraser University has been involved in funding these patent applications and developing applications for external funding.

During the ninth quarter initial steps were taken toward establishing a private corporation based on developing implantable technologies.  The founding members of this corporation are Andy Hoffer, Klaus Kallesøe, Kevin Strange, and Ignacio Valenzuela, and the intended name for the entity is Neurostream Technologies Inc. which has been reserved with the British Columbia Government.

## E.    Reporting results at meetings

During the ninth quarter two papers emerging from this contracted research were accepted for publication in the proceedings of the Rehabilitation Society of North America meeting in Vancouver, Canada, June 1995.  The first paper, authored by Kevin Strange et al., is entitled "Long term stability of nerve cuffs implanted in the cat forelimb" and presents CAP data from the two series of NIH implants.  The second paper, authored by Paul Christensen and Andy Hoffer, is entitled "An eight-channel biphasic stimulator for functional electrical stimulation" and details the development of a stimulator designed for use with the NIH implants in Year Three.  These two papers are included in this report.

In addition, an extended abstract authored by Kevin Strange and Andy Hoffer entitled "Using cutaneous neural signals to predict muscle activity during walking in the cat forelimb" has been submitted for publication in the proceedings of the IEEE Systems, Man, and Cybernetics meeting in Vancouver, Canada, Oct. 1995.

# ᴵV. Plans for Tenth Quarter

In the tenth quarter we intend to:

1. examine histopathologically the nerves from Year One and Year Two cats (objective 5)

2. complete the study of two remaining Year Two cats (objective 4)

3. design and construct cuffs appropriate for smaller proprioceptive nerves (objective 3)

4. complete the construction of an 8-channel stimulator to be used for FES of forelimb muscles (objective 4b)

5. complete the construction of hardware and begin the software design for controlling the reaching task (objective 4a,b)

6. develop a model of closed loop control of FES during walking utilizing neural feedback (objective 4)

7. analyze walking data with our collaborators (objective 7)

# V.  References

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# VI. Attachments

1. Strange, K.D., Hoffer, J.A., Kallesøe, K., Schindler, S.M., and Crouch, D.A.  Long term stability of nerve cuffs implanted in the cat forelimb. in *Proc. of RESNA*, Vancouver, In Press.

2. Christensen, P.R. and Hoffer, J.A.  An eight-channel biphasic stimulator for functional electrical stimulation. in *Proc. of RESNA*, Vancouver, In Press.

# LONG TERM STABILITY OF NERVE CUFFS IMPLANTED IN THE CAT FORELIMB

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## ABSTRACT

In two series of experiments, we chronically implanted a total of 56 nerve cuffs in the forelimbs of 14 cats. Each nerve was instrumented with a proximal stimulating cuff and a distal recording cuff, and we periodically monitored the status of the nerves and implanted devices over a total implant period of six months by stimulating the nerve and analyzing the resulting compound action potentials (CAPs) in the distal cuffs. Improvement in CAP amplitude and latency stability from the first series of implants to the second is demonstrated, as a result of improvements in surgical skill, quality of devices, and protection of hardware external to the cat. The longevity of neural recordings and implanted devices supports their implementation in human applications of functional electrical stimulation such as neural prostheses.

## BACKGROUND

Recent advances in functional electrical stimulation (FES) systems have led to the consensus that closed-loop control of FES is both desirable and necessary for returning more efficient and normal-like function to paralyzed extremities. External sensors can be used to provide force and position feedback for neural prostheses systems, yet external devices are often inaccurate and difficult to position in the environment without external braces. An attractive approach is to utilize the natural sensors available in the extremities to retrieve contact, force, and position information from peripheral nerve and muscle recordings [1,2]. Recently, a novel clinical application of a closed-loop FES system to correct for drop-foot in a stroke patient implemented the use of cutaneous activity recorded from the Sural nerve with a chronically implanted nerve cuff as feedback [3]. This group has now also implanted a recording cuff on a human Palmar Digital nerve to retrieve contact and force information from the index finger

(demonstrated at the Neural Prostheses: Motor Systems IV conference in Ohio, July 1994).

## OBJECTIVES

Our overall objective is to identify reliable sources of neural control signals and extract features, such as ground contact and slip information, that may be applied in closed-loop FES systems for restoring voluntary use of paralyzed muscles in humans [4,5]. We are investigating both skin contact and limb position information obtained from nerve cuffs implanted on cutaneous and proprioceptive nerves in the cat forelimb during behavioural tasks.

Our objective for this study was to determine the stability of forelimb nerves of cats that were chronically implanted with stimulating and recording nerve cuffs over periods of six months.

## METHODS

In a first series of implants, we implanted four nerve cuffs in the left forelimb of each of eight cats, two on the Ulnar nerve and two on the Median nerve, as shown in Fig. 1. Each nerve was instrumented with a proximal stimulating cuff above the elbow and a distal recording cuff below the elbow. We also implanted ground wires, a thermistor to record local limb temperature, and electromyogram (EMG) electrodes to record muscle activity in the Palmaris Longus, a primary wrist flexor. The wires were routed to a common percutaneous exit point and attached to a specially designed backpack containing a printed circuit board and a 40-pin connector.

In a second series of experiments, we implanted four cuffs in the left forelimb of six further cats, instrumenting two of the Ulnar, Median, and/or Radial nerves. In this series we also implanted EMG electrodes in two wrist flexors, the Palmaris Longus and the Flexor Carpi Ulnaris, and two wrist extensors, the Extensor Carpi Ulnaris, and the Abductor Pollicis Longus.

Our research animals were group-housed in a room that includes climbing shelves up to 2 m high and

were freely allowed to jump and play. In order to simulate an implant in an active person, we did not attempt to control the physical duress applied to the implanted devices by restricting the cats' movement.
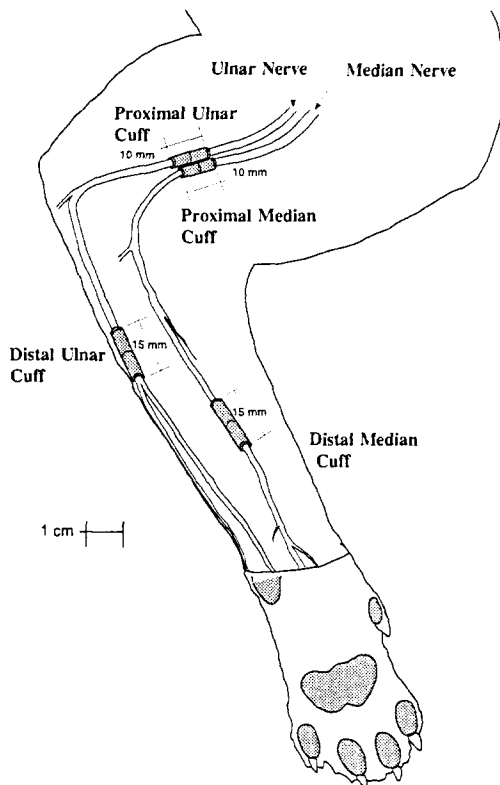


Figure 1: Medial view of the cat left forelimb with implanted nerve cuffs

Under halothane gas anesthesia, we periodically monitored the status of the nerves and implanted cuffs by stimulating the proximal nerve cuffs at supramaximal current levels and recording the compound action potentials resulting in the distal cuffs. We monitored the peak-peak amplitude of the CAP and the latency or conduction time between cuffs, both of which are indicative of the health of the instrumented nerve and viability of the nerve-cuff interface. We also monitored cuff electrode impedance and EMG pickup and recorded nerve and muscle data in the awake cat performing behavioural tasks.

Our surgical and experimental protocols were in accordance with guidelines set by the Canadian Council of Animal Care and were approved by the University Animal Ethics Committee.
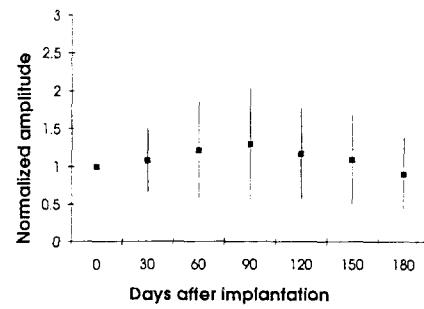
RESULTS



Figure 2: CAP amplitudes from first series of implants (normalized at day 0; n = 9; cats 2-6,8)
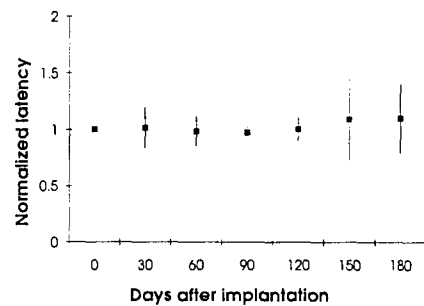


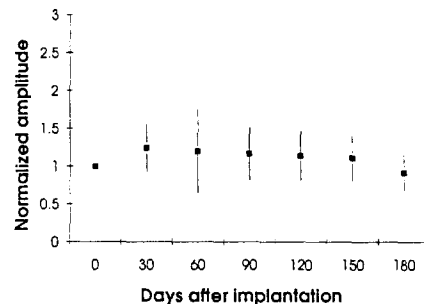Figure 3: CAP latencies from first series of implants (normalized at day 0; n = 9; cats 2-6,8)



Figure 4: CAP amplitudes from second series of implants (normalized at day 0; n = 10; cats 9-14)
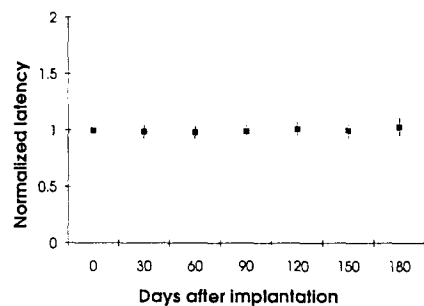


Figure 5: CAP latencies from second series of implants (normalized at day 0; n = 10; cats 9-14)

Figures 2 and 3 show the averaged CAP amplitude and latency data from the first series of cats. The CAP amplitude and latency data for each nerve monitored for longer than 180 days were interpolated to sampling intervals of 30 days. In the first series of implants, we successfully recorded from 9 nerves past 180 days, with the 7 failures resulting from pulled wires at the percutaneous exit point (5) and damage to the nerve (2). Both the average amplitude and latency data were stable throughout the implant period with maximum standard deviations of 57% (at day 90) and 33% (at day 150) respectively.

Figures 4 and 5 present the average CAP amplitude and latency data from the second series of implants. We recorded successfully from 10 nerves, with the two failures resulting from one case of pulled wires and a nerve compression injury. In the second series, the CAP amplitudes and latencies were even more stable than in the first series, with maximum standard deviations of 46% (at day 60) and 8% (at day 180) respectively.

DISCUSSION

Our results show that recording nerve cuffs can be implanted in the forelimbs of unrestrained cats for periods of at least six months without inflicting damage to the nerve, providing that 1) sufficient care is taken when surgically manipulating the nerve and installing the cuff, and 2) provisions are taken to prevent the cats from pulling and breaking wires. We attribute our greater success in the second series to both of these factors as we implemented an improved nerve cuff closing method which reduced trauma during surgery, and a fabric band around the belly of the cat which restricted access to the backpack connector and wires.

Future improvements in implantable nerve cuff designs in terms of mechanical interaction with the nerve and development of superior electrode materials and designs may improve the quality of neural recordings and longevity of recording nerve cuff applications. A further improvement in implantable device systems would be to utilize telemetry systems to avoid wire breakage inside the body or at percutaneous connectors.

Our demonstration of the stability and longevity of neural recordings in the cat forelimb lends support to including neural feedback in human applications of FES systems and neural prostheses.

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# AN EIGHT-CHANNEL BIPHASIC STIMULATOR FOR FUNCTIONAL ELECTRICAL STIMULATION

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## ABSTRACT

Fluid limb motion control produced with functional electrical stimulation (FES) requires accurate timing of current pulses independently delivered to various muscles as well as the ability to rapidly change the intensity and patterns of stimulation in response to changes in load, limb properties, or command signals [1]. In our laboratory we are studying ways to use feedback information contained in neural activity during active movements with the goal of restoring the voluntary use of paralyzed limbs. The stimulator that we have designed is an integral part of the research being conducted to mimic normal movement control. This paper presents the criteria that we employed to design an eight-channel, feedback-controlled biphasic stimulator. The design is a novel approach which incorporates digitally programmable resistors to control the current amplitude of the delivered waveforms.

## BACKGROUND

The goal of a constant current stimulator is to deliver accurate and reliable current pulses to muscle or nerve electrodes so that the net delivered charge is known. By varying the amount of charge delivered, the number of motor axons recruited by a stimulus pulse can be controlled in order to change the muscle force. Waveforms with amplitudes in the milliamp range are necessary to stimulate intramuscularly, whereas currents in the microamp range are necessary to stimulate nerves directly.

A stimulator with multiple output channels is needed in order to program stimulation patterns that resemble the normal physiological stimulation patterns in muscles of interest. Individual muscles need to be stimulated at different frequencies to produce the desired joint torques. By using biphasic pulse stimulation, the charge delivered to the stimulating electrode can be balanced thus reducing toxic reactions at the tissue-electrode interface [2, 3]. The first pulse triggers an action potential within the axon, while a second pulse of opposite polarity, smaller amplitude, and longer duration balances the delivered charge.

The stimulator that we have designed can stimulate up to eight muscles or nerves simultaneously and allows independent on-line control of the width of every delivered current pulse.

## STATEMENT OF THE PROBLEM

The stimulator that we required needed to be able to synchronize up to eight stimulation patterns and interchannel intervals and accurately produce the biphasic pulses needed for stimulation. The stimulator needed to deliver positive and negative polarity (i.e., biphasic) rectangular-shaped current pulses in the range 10 $\mu$A to 30 mA through a maximum impedance of about 15 M$\Omega$. Successive pulses could have the same or opposite polarity, but would usually consist of a positive pulse followed by a negative pulse. Each channel was to be stimulated sequentially with a nominal frequency of 25 Hz per channel with pulse widths up to 2550 $\mu$ sec. The stimulator also needed to be flexible enough so that several users could apply it to their own specific purposes easily and quickly.

The new stimulator was to be programmable for use in a closed-loop configuration where the program could modify any of the stimulation parameters (frequency, pulse duration, pulse amplitude) once a control paradigm was determined. As an example, this paradigm may take the form of a state machine combined with a threshold function. If the recorded electromyogram (EMG), electroneurogram (ENG), or muscle force suggests that the limb is in some defined state, then the stimulator will change its output to make the limb take appropriate action. When the action is not appropriate, then the stimulator's output will be modified on the basis of ongoing feedback [1].

## APPROACH

To enable a flexible system, we have utilized a 486 computer as the controller for the stimulator. The computer controls the channel sequencing and timing and operates as the user interface. From the user's point of view, the stimulator is comprised of

a computer connected to the stimulator hardware (see Figure 1). The computer provides a mouse-driven interface that allows the user to select the number of channels to stimulate and to control the stimulation parameters.

We adopted a modular design which simplifies the realization of the circuitry. Each of the counter blocks comprising the timing circuitry has the same structure and all the current sources have the same design and layout. If any particular section of the stimulator board needs to be modified, then that section can be removed and replaced. For example, each current source is built on a separate card so that current sources may be easily replaced if this should become necessary.

## SYSTEM DESIGN

The hardware section of the stimulator can be divided into the controlling computer, the interface and decoding circuitry, the timing circuitry, and the current source circuitry. The block diagram in Figure 1 shows the interconnections between the various sections.

The computer is responsible for the overall control of the system and the user interface. We programmed a 486 computer to act as the controlling computer and wrote routines in C with Borland's C++ 3.1 compiler to communicate with the stimulator hardware. The user interface for the stimulator was based upon an existing interface used previously in the lab [1]. The computer case also houses an Omega PC-CTR-20 Counter/Timer board that handles the gross timing between channels and a National Instruments AT-MIO-16F-5 acquisition board that interfaces with the stimulator board.

The interface circuitry receives four-bit data nibbles from the computer and stores them in latches and therefore acts as a virtual 32-bit wide control port: 16 bits of this port are reserved for control signals and the other 16 are reserved for load bytes to the counter chips to allow for fast loading into the counters. The 16 control bits are divided to select one of the eight channels, to control one of the four programmable resistors on the selected current source, to load the counters in the timer circuitry, and to set the polarity and range of the current pulses.

The decoding circuitry selects a current source for stimulation and selects a programmable resistor in one of the current sources to set the current amplitude.
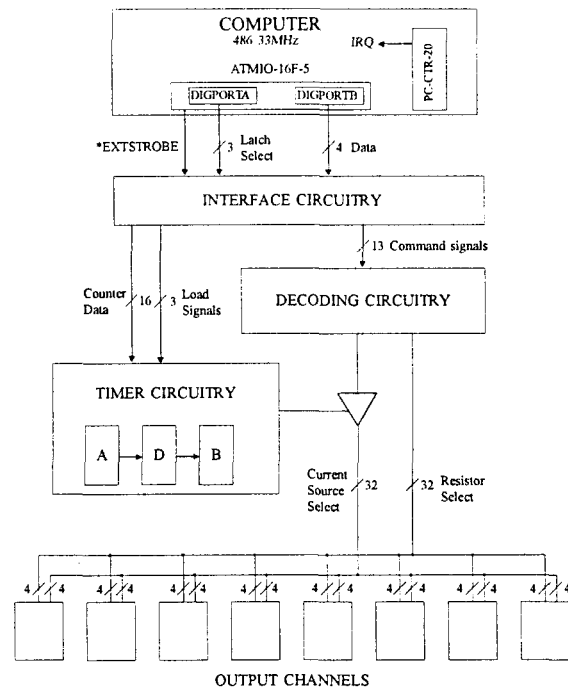


Figure 1: Stimulator block diagram

The timing circuitry controls the precise timing of the activation of the current sources to the nearest microsecond. When counter blocks "A" and "B" are active, they enable the command signal to the current source selected by the decoding circuitry. When these counters have finished counting or when the delay counter (counter D) is active, the command signals to the current sources are disabled.

The optically isolated current sources are located at the final output stage of the stimulator board and are connected to electrodes which stimulate nerves or muscles. When the command signal is asserted, the current source turns on producing a rectangular-shaped pulse that ends when the signal is deasserted. Four current sources compose each channel. Two current sources are necessary to obtain the full $3^{1}/_{2}$ decades of range of currents because the digitally programmable resistor itself (Xicor $E^2POT$ X9C103P) only has 2 decades of range. One is designed to handle 10 µA to 1,000 µA in 10 µA steps, whereas the other is designed to handle 300 µA to 30 mA in 300 µA steps. We duplicated this pair of current sources to get the opposite polarity current pulse.

## DISCUSSION

With digital control, the user knows the exact state of the stimulator. A configuration file may be created that stores a user's setup of the stimulator parameters (e.g., number of stimulating channels, pulse amplitudes and durations) and can be reloaded at a later time. The digitally programmable resistor values are known precisely and can be reconfigured identically each time the stimulator is used.

The stimulator has the ability to respond and adapt to incoming feedback signals. Pulse widths, amplitudes, and frequency of stimulation can all be controlled to ensure that the stimulated limb takes appropriate action.

With our stimulator, a user has the ability to independently control up to eight muscles. If the user's objective is to control movement about a joint, a software program will cause the stimulator to alternate the stimulation of antagonist muscle groups. Alternately, by stimulating both muscle groups simultaneously, antagonist muscle groups could be coactivated in order to increase the stiffness of the joint.

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